Sex differences in dynamic knee stability in high intensity functional training athletes.

by

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Abstract

INTRODUCTION. Acute anterior cruciate ligament (ACL) injury is the single greatest concern in modern orthopedic sports medicine. While sex is a known predictor for knee injury among sporting populations, relative risks for knee injuries among men and women in high intensity functional training (HIFT), an activity that blends sport and fitness together, are unknown. Further investigation into the fatigue-related compensation mechanisms and dynamic knee stability of HIFT-trained athletes is necessary to understanding the comparative risk of participation in these exercise modalities. **METHODS.** Participants completed a single-group 2-visit assessment protocol consisting of a familiarization session in which anthropometric data were collected, as well as a testing session. During the testing session, participants completed a three-round HIFT workout, as well as a biomechanical performance analysis utilizing the single-leg squat and drop vertical jump before and after each round of the workout. **RESULTS.** No significant differences existed in mean round times between sexes (F=.53, p=.48, η^2=.03), as well no significant main effect for time (F=3.60, p=.07, η^2=.17) or interactions between sex and time were detected (F=.01, p=.95, η^2=0). There was a significant main effect of time on heart rate (F=213.31, p<.001, η^2=.92) and rate of perceived exertion (F=22.75, p<.001, η^2=.56), but no interaction between sex or sex*time existed for either variable. No significant multivariate effects for time or interaction between sex*time were present for stance time or jump height. There was a significant multivariate main effect for sex on baseline drop vertical jump and right single-leg squat kinematics and kinetics, but not for the left single-leg squat (F=1.29, p=.26, η^2=.26). These sex differences during the drop jump and right single-leg squat existed across all time points, but no significant multivariate main effects for time (drop vertical jump: right leg: F=1.11, p=.31, η^2=.06; left leg: F=1.7 p<.01, η^2=.09; Right single-leg squat: F=.78, p=.83, η^2=.05) or sex*time (drop vertical jump: right leg: F=.76, p=.85, η^2=.04; left leg: F=1.08, p=.34, η^2=.06; Right single-leg squat: F=1.08, p=.35, η^2=.04) were found between groups. No multivariate main effects for time (F=.70, p=.91, η^2=.04) or interaction between sex and time (F=0.60, p=.97, η^2=.03) existed during the left single-leg squat. **DISCUSSION.** While participants perceived themselves be working harder as the workout progressed, results did not suggest fatigue was present. No differences were detected at baseline for the left single-leg squat. Conversely, right single-leg squat baseline data suggested that women have less dynamic control of their dominant limb than men. While sex differences existed across all timepoints for both movements, time and sex*time were not predictors for deterioration in knee mechanics. **CONCLUSION.** Further, while sex differences existed in movement mechanics, they did not deteriorate throughout the workout, suggesting that HIFT may not induce levels of fatigue significant enough to compromise movement mechanics. This study also suggests that HIFT trained women’s movement mechanics appeared superior to what has been previously reported in the literature.

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Chapter 1 - Introduction

Knee Injury Epidemiology in Sport

Sports and recreational activity-related injuries hold a place among society's major public health problems due to the social and economic burden they present (Belechri et al., 2001; Öztürk & Kiliç, 2013). In an epidemiological analysis of 16 years of injury data across 15 men's and women's collegiate sports, more than 50% of all reported injuries to athletes were to the lower extremities (Hootman et al., 2007). Specifically, the most common location for injuries to the lower extremity is the knee (29%) (Darrow et al., 2009; Taunton et al., 2002).

From 1999 to 2008, an estimated 6,664,324 (average of 666,432 per year) sport-related knee injuries were presented to emergency departments across the United States (Gage et al., 2012), with acute anterior cruciate ligament (ACL) injury continuing to be the single greatest problem in orthopedic sports medicine (Renstrom et al., 2008). Between 100,000 to 200,000 ACL ruptures occur each year in the United States alone (Gordon & Steiner, 2004), and ACL injury rates have continued to increase over a 15-year time period (Hootman et al., 2007). A large percentage of these ACL injuries result from non-contact related athletic exposures; a term the NCAA reports as quantifying a game, practice, or training session completed by an athlete (Hootman et al., 2007).

Members of the military, firefighting, and police forces, often labeled as tactical athletes, have demonstrated similar injury trends as sport populations (Gwinn et al., 2002). Among these tactical athletes, women had a 1.4 to 11.9 times greater relative risk for non-contact ACL injuries than men (Gwinn et al., 2002). While ACL tear incidence rates in the general population are approximately 68.6 injuries per 100,000 people per year, actual incidence rates may be higher due to the lack of a standard surveillance system for recording injuries (Sanders et al., 2016).

Women also experience more ACL injuries per athletic exposure across all sports than men (Agel et al., 2016; Dick et al., 2009). Darrow and associates (2009) reported on the epidemiology of severe injuries among high school athletes, and cited women as consistently experiencing higher rates of severe knee injuries in comparable sports (i.e. soccer, basketball) than men. Further, ACL injuries account for a greater proportion of total team injuries in women's sports (soccer, lacrosse, gymnastics, and basketball) than equivalent men's sports (Renstrom et al., 2008). Further, when considering injuries per 1000 athletic exposures, which the NCAA reports as one session of sport (either practice or play) rather than in terms of hours of exposure, women's gymnastics yields the highest incidence rates among all men's and women's sports (Renstrom et al., 2008). While these sex differences are clearly known predictors within sporting activities, differences between men and women in high intensity functional training (HIFT), an activity that blends sport and fitness together, are unknown.

HIFT: Bridging Sport and Fitness

HIFT is a modality of exercise that emphasizes constantly varied, functional, multi-joint movements that may be modified to any fitness level, which are often performed at relative high intensities (Feito, Heinrich, Butcher, & Poston; 2018). HIFT has been popularly branded under Greg Glassman and Lauren Jenai's CrossFit®, which, since opening its first affiliate in 2001, has become the largest fitness franchise in the world (Feito et al., 2018). CrossFit®'s explosive growth over the past two decades has generated massive increase in general population participation as the fitness franchise has branded itself the "Sport of Fitness." Athletes around the world now compete locally as well as on the global scale with competitions such as The CrossFit® Open, Granite Games, Rogue Invitational, and The CrossFit® Games.

HIFT-style programs, including CrossFit® have received criticism from those in academia, medical professions, and popular media (Drum et al., 2016; Greely, 2014; Powers et al., 2014) on the basis that they present an increased risk for bodily harm to participants (Katz et al., 2016). This is primarily due to assertion that because HIFT exercises are technically demanding and require sustained levels of high power output, participants are likely to experience significant muscular fatigue (Maté-Muñoz et al., 2017).

Studies have reported that HIFT has the demonstrated ability to cause muscular fatigue (Maté-Muñoz et al., 2017), and that fatigue results in modifying movement biodynamics (Weisenthal, Beck, Maloney, DeHaven, & Giordano, 2014). However, these studies have focused on individual modalities of HIFT workouts such as aerobic conditioning, gymnastics, or weightlifting to determine whether muscular fatigue was occurring. While these modalities are acceptable to study individually, HIFT workouts are often highly integrated and involve performing movements from multiple modalities in workouts lasting from 3-20 minutes in length (Feito, Patel, Sal Radondo, & Heinrich, 2019). From a methodological standpoint, HIFT is built on the foundation that fitness is best achieved through successful interaction between the body's high power anaerobic and low power aerobic energy systems (Glassman, 2002). This concept is exemplified in most HIFT workouts which last from 6-12 minutes to allow for high intensities to be maintained throughout.

Although fatigue has been shown to alter movement mechanics in controlled conditions (Thomas, McLean, & Palmieri-Smith, 2019), the notion that HIFT is dangerous is largely refuted by the growing body of evidence that HIFT training results in injury rates similar to modalities including Olympic weightlifting, basic weightlifting, gymnastics, and a multitude of other sports that require similar metabolic and physical demands(Moran et al., 2017; Poston et al., 2016;

Raske & Norlin, 2002). Overall, the reported retrospective injury rates of HIFT participants range from .27 to 3.2 per 1,000 hours trained (da Costa et al., 2019; Feito, Burrows, & Tabb, 2018; Hak, Hodzovic, & Hickey, 2013; Moran et al., 2017). Injuries among HIFT participants most frequently occur at the knee, along with shoulder and low back (Feito, Burrows, & Tabb, 2018; Weisenthal, Beck, Maloney, DeHaven, & Giordano, 2014). Many non-contact sports elicit similar physiological stress on the body (Vanheest, 2008) as HIFT, further supporting that investigation into the dynamic knee stability patterns of HIFT-trained athletes is necessary to understanding the comparative risk between these exercise modalities.

Movement Efficiency

Studying knee stability requires biomechanics research. Simply stated, biomechanics seeks to describe the relationship between force applied to a specific tissue and that tissue's subsequent ability to tolerate that force, which determines the outcome of the given interaction (Bartlett & Bussey, 2013). Sport-related biomechanics can be thought of as having two overarching goals, which are to increase performance while also reducing risk for injury (Lynn & Noffal, 2012).

Biomechanical efficiency, as it relates to human movement, describes patterns of movement that fulfill their tasks with minimal strain on the musculoskeletal system (Pitt-Brooke et al., 1998). This classical definition can also be expanded to include the ability to perform movement without pain or discomfort and involves proper joint alignment, muscle coordination, and posture (Kritz et al., 2009). More efficient movement patterns have the ability to increase performance, while potentially reducing the risk for injury because of the reduced strain placed on the body (Sahrmann, 2002).

The importance of this biomechanical methodology for improving performance and reducing risk for injury is exemplified when applied directly to sport settings. In this way, sports are two-dimensional in that while they offer potential health benefits, they also expose one to the risk of injury. A positive correlation exists between athletic exposures and injuries, which places any person who participates in sports at risk proportionate to their participation level (Öztürk & Kiliç, 2013).

Biomechanical Basis of Dynamic Knee Stability

In order to understand hat biomechanical efficiency looks like in an applied setting, it is appropriate to first gain a general understanding of knee joint anatomy and the general mechanics of each of its components that contribute to knee stability. The knee is a modified hinge joint that functions to allow flexion and rotation while also maintaining complete stability and control under a large range of loading conditions (Buckwalter et al., 2000). The knee consists of the patellofemoral joint and the femorotibial joint, which stabilize the knee by working in conjunction with the bony architecture and static and dynamic restraints of the ligaments, capsule, and musculature that cross the joint (Goldblatt & Richmond, 2003).

The femorotibial joint is comprised of a medial and lateral condyle, each of which articulates with the corresponding tibial plateaus. Medial and lateral menisci sit between these two articulations, enhance the conformity of the joint and assist with rotation of the knee. It is important to note that the lateral condyle of the femur is smaller than the medial condyle, which is a contributing factor in the naturally occurring valgus and anteroposterior alignment of the knee (Goldblatt & Richmond, 2003). Knee valgus is a condition in which the angle formed at the knee between the femur and tibia angulates towards the midline with concurrent tibial angulation away from the midline, and high ground reaction forces (Andrews & Axe, 1985). The patellofemoral joint is a sellar joint which sits on the anterior aspect of the knee and serves as the point of interaction between the patella and femur (Grelsamer & Klein, 1998). The patellofemoral joint plays an important role in stabilizing the knee through the extensor mechanism (Buckwalter et al., 2000).

Knee stability is controlled by a variety of factors including ligament and soft-tissue restraints, active muscular control, condylar anatomy, and tibiofemoral contact forces at the joint interface during weight-bearing activities (Markolf et al., 1981). Recent evidence suggests that biomechanical or neuromuscular imbalances of the lower limbs during dynamic movement could be a primary contributor to ACL injuries, especially in women (Boden et al., 2000; Zazulak et al., 2008). Neuromuscular deficits related to dynamic knee stability and coordination are ligament dominance, quadriceps dominance, and leg dominance (Hewett et al., 2001).

First introduced by Andrews & Axe (1985), ligament dominance occurs when lower extremity musculature does not adequately absorb the forces exerted by an athletic movement resulting in excessive loading of the knee ligaments. This is especially true of the ACL, which plays a key role in resisting knee valgus and anterior tibial translation. This lends further support to the importance of proper activation of posterior chain musculature for proper absorption of forces about the knee joint (Andrews and Axe, 1985).

Quadriceps dominance describes an imbalance in the muscle recruitment patterns of the knee flexors and extensors (Hewett et al., 1996). Specifically, this imbalance refers to the tendency to preferentially knee extensor moments over knee flexor moments when performing athletic movements that generate significant lower extremity joint torques (Hewett et al., 1996). Women tend to rely heavily on the quadriceps over the hamstrings to stabilize the knee joint during dynamic movements such as landing and jumping (Ford et al., 2003), which causes an increase in anterior shear forces and ground reaction forces, thus increasing the load on the ACL. Further, landings that load the quadriceps primarily result in more erect truck orientations. This poses a problem as it shifts the body's center of mass more posteriorly, increasing the lever arm for ground reaction forces at the knee and resulting in greatly increased external knee flexion moments (Dingenen et al., 2015). These aggressive quadriceps loading mechanics have been shown to induce non-contact ACL injury (Decker et al., 2003; DeMorat et al., 2004; Huston et al., 2001; Krosshaug et al., 2007).

Limb dominance is simply an imbalance in muscular strength and recruitment patterns between the body's two lower limbs, where one side demonstrates greater control over dynamic movements than the other (K. R. Ford et al., 2003). The injury risk associated with limb dominance is two-fold. The first issue is that an athlete's dominant leg is overused and experiences elevated levels of joint torques. The second issue is that the non-dominant limb may not possess the musculature to effectively absorb the high forces exerted on it during dynamic movement (Myer, Ford, & Hewett, 2004).

Finally, age is an important consideration when assessing differences in biomechanics between populations. The literature supports the notion that the hip, knee, and ankle biomechanics deteriorate with increasing age (Alexander et al., 1991; DeVita et al., 2016). However, no current research offers justification as to a specific point in the human lifespan when these changes can be anticipated. Athletic performance declines in a linear fashion throughout the lifespan, but peak performance ages have been calculated anywhere from 25-41 years of age depending on the modality of exercise (Ganse et al., 2018). Just as well, a variety of factors including genetics, comorbidities, and lifestyle greatly influence these declines in athletic performance.

Influences of Fatigue on Knee Joint Kinematics

In addition to the influences of sex on dynamic knee stabilizing mechanisms, fatigue is also thought to affect ACL injury risk. Previous research has suggested fatiguing close kinetic chain exercise protocols, which involve movements where the end of the "chain" furthest from the body would be anchored in place such as the feet during a squat, result in compromised neuromuscular strategies for controlling the knee joint during dynamic landing movements (Gehring et al., 2009). Primarily, this refers to a reduced pre-activation of the medial and lateral hamstrings and gastrocnemius muscles in both men and women, which causes elevated levels of stress to the knee joint (Gehring et al., 2009).

Similar studies found the post-fatigue landing characteristics of both men and women to be consistent with non-contact ACL injury mechanisms. Women presented with a greater post-fatigue effect in knee anterior shear force and less knee flexion than men (Kernozek et al., 2008). This, along with a demonstrated inability in women to generate knee varus moments similar to men at peak knee valgus position (Kernozek et al., 2005), causes the knee joint to deviate laterally toward the midline. This may place women at an even greater predisposition to ACL injury when exposed to highly fatiguing exercise and movements that require repetative high impact knee stabilization.

While studies have used drop landing and vertical jump protocols to assess the impact of fatigue on knee joint kinetics and kinematics (Mejane et al., 2019; Weeks et al., 2015), we identified no current literature employing the use and measurement of the drop vertical jump as well as the single-leg squat under the same potential fatiguing conditions over multiple time points. No consensus exists within the scientific community as to whether women are more or less prone to the effects of fatigue on lower extremity biomechanics. What we can be sure of is that HIFT builds its foundation on emphasizing movement efficiency and fidelity above all else in its participants.

While the biomechanics literature has linked these potential fatigue-related mechanisms to increased injury risk (Kernozek et al., 2005, 2008), muscle physiology literature has claimed that women actually possess an advantage over men in fatigue resistance, especially during protocols which incorporate submaximal protocols of 20-70% of maximum voluntary contraction (MVC) (Fulco et al., 1999; Lindstrom et al., 1997). Multiple mechanisms have been identified as potential explanations for this fatigue resistance.

For one, the relatively lower absolute muscle forces generated by women for the same relative work as men (Hicks et al., 2001) should result in less oxygen demand and mechanical compression of local vasculature (Maughan et al., 1986). Additionally, differences in substrate utilization (Tarnopolsky, 1999) and neuromuscular activation patterns (Hakkinen, 1993) have also been identified as potential mechanisms which differentiate women from men. It is important to note, however, that women's reported fatigue-resistance superiority declines with muscle contraction intensity (>80% MVC) (Maughan et al., 1986). HIFT workouts are generally performed at relative high intensities, which lends support to the notion that little difference in response to fatigue should be expected.

Human Movement Analysis Techniques

Current technology allows 3-D joint kinetics and kinematics to be captured either using biomarker or markerless-based motion capture systems. Marker-based motion capture systems have traditionally been the "gold standard" for human biomechanical research as they allowed for the development of pre-screening techniques for identifying landing kinematics that may predispose athletes to ACL injury, but have been challenged by the introduction of markerless-based systems which can produce the same results while eliminating multiple sources of error (Kohler, 2012). Both systems measure joint kinematics on a sub-millimeter level (Perrott et al., 2017), and the minuscule differences in the accuracy of these systems has made it impossible for scientific community determine which system actually measures movement more accurately and produces less error (Ceseracciu et al., 2014). What separates markerless technology from its marker-based counterpart is its repeatability, feasibility, and clinical relevance.

Marker-based motion capture systems rely on humans for placement of markers for each subsequent testing session, which has the potential for high amounts of error when placing upwards of 50 markers on a subject each testing session (Fuller et al., 1997). Additionally, because markers are placed on soft tissue, they have the propensity to move while the body is in motion, which may confound results and is far less repeatable than markerless systems (Reinschmidt et al., 1997; Sati et al., 1996). Finally, because of the extensive time required for marker placement, marker-based systems are structurally restricted from keeping up with the efficiency of markerless systems, which can see from 50-100 people each day (Corazza et al., 2006). Thus, markerless methods of data collection eliminate these errors and inefficiencies.

Specific Aims

The first aim of this study was to determine whether a workout incorporating traditional HIFT movement modalities and appropriate time standards (Glassman, 2002) caused significant fatigue in HIFT-experienced men and women. We hypothesized the workout would cause significant fatigue in both men and women. Further, we predicted that men and women would both display similar objective (i.e., heart rate) and perceived (i.e., rating of perceived exertion) responses to the workout.

The second aim of this study was to determine whether sex differences in movement mechanics existed at baseline. In agreeance with previous literature (F. R. Ford et al., 2003; Kernozek et al., 2005, 2008), we hypothesized that women would exhibit movement patterns more consistent with known predictors for non-contact ACL injury than men at baseline.

The third aim of this study was to determine whether sex was associated with changes in dynamic knee stability across the workout. We hypothesized that there would be no differences by sex for changes in dynamic knee stability across time. HIFT emphasizes a demonstrated proficiency in foundational movements as critically important prior to placing an athlete under any sort of load or increased intensities. Many gyms implement mandatory on-ramp programs for new clients where they receive individual instruction on foundational movement patterns before joining regular group classes. Further, coaches provide constant support during workouts in order to ensure safety and fidelity of movement. Thus, although we expected women to demonstrate less knee stability than men at baseline, we did not expect their knee stability to deteriorate significantly more during the workout protocol. This would suggest a link between participation in HIFT and more efficient movement mechanics under fatigued states, which has clear injury prevention implications.

Chapter 2 - Methods

Design

This study used a single-group 2-visit assessment protocol. Based on the magnitude of the interaction effect between fatigue and lower extremity joint mechanics, priori power analysis (G*Power v. 3.1.9.2, Universität Kiel, Germany) indicated that a sample of 20 participants across both genders was needed to achieve B of .80 at an Alpha of .05 at 80% power. All study procedures and protocols were approved by Kansas State University's Institutional Review Board (IRB approval #9793).

Participants

Participants included 10 men and 10 women between 18-35 years of age who had been participating in HIFT at least three hours per week for the past six months. To be included in the study participants had to be able to perform the single-leg squat and drop vertical jump (as described below) unassisted and unmodified. Participants also had to demonstrate proficiency for the workout movements of dumbbell thrusters, air squats, and burpee box jumps. Exclusion criteria included explicitly training for a specific performance outcome (e.g., marathon, triathlon, powerlifting or weightlifting competition, bodybuilding show, etc.), history of ACL injury or diagnosed knee pathology by a physician, lower limb fracture or surgery, or lower body injury within the three months prior to participation. Rolling recruitment of participants was utilized over a 6-week period. Participants were recruited in-person via announcements prior to classes at HIFT-based gyms in Manhattan, Kansas, as well as via social media accounts associated with our research laboratory, department, and college. Fliers were also posted at local HIFT gyms.

Measures

Anthropometrics. Participant height was measured using Charder HM200P Stadiometer (Issaquah, Washington, USA), and body weight, body mass index (BMI), body fat percentage (%), fat mass, and fat free mass were recorded using bioelectrical impedance via a Tanita TBF-310 (Arlington Heights, Illinois, USA).

Biomechanical data. Biomechanical data were collected using a validated 3D marker less motion capture system from the Dynamic Athletic Research Institute (DARI), which utilizes eight-cameras to acquire a skeletal model of the body that measures knee biomechanics during dynamic movement. Hip and knee joint kinematics (i.e., positional change) and kinetics (i.e., moments) were assessed during both the single-leg squat and drop jump movement tasks. Prior to each use of the DARI system, manufacturer-provided calibration techniques were performed to protect both validity and reliability of the kinematic and kinetic data collected. Movement order was randomized for each trial to ensure that potential error was spread evenly throughout the movements.

Joint Kinematics. Participants' joint kinematics were assessed similarly in the single-leg squat and the drop vertical jump. During the single-leg squat, data including hip adduction/abduction angles, minimum and maximum knee valgus, dynamic knee valgus, knee valgus at a low point of the squat, hip adduction/abduction at low-point, and peak knee flexion angles were collected. Regarding the drop vertical jump, hip adduction/abduction, knee and hip flexion, and knee valgus were assessed at contact. These measures were also assessed at the maximum loading timepoint, along with minimum, maximum, and dynamic knee valgus throughout the movement. For data interpretation of knee varus/valgus and hip abduction/adduction, deviation towards the midline was denoted as a positive angular value, and deviation laterally was negative.

Joint Kinetics. Joint kinetics were assessed during the drop vertical jump and expressed in Newtons; the standard unit for force. Ground reaction force (GRF) data were collected in time-series fashion and normalized for each timepoint of interest during the movement. GRF data were collected from initial contact to takeoff during the jump. Knee joint torques were also assessed at peak loading and takeoff.

Fatigue Parameters. Training HR was assessed via a commercially available continuous heart rate monitor (Polar H7® heart rate monitor, Polar Electro, Inc., Bethpage, New York, USA). Participants were fitted with the heart rate monitor strapped across the chest with the monitor centered at the distal end of the sternum. Heart rate data recording began immediately upon commencement of the first DARI analysis. Participants reported their rate of perceived exertion (RPE) immediately following completion of each round of the workout. RPE was assessed using the original Borg's RPE scale (i.e., 6-20; 6 being no exertion or very light and 20 being maximal exertion or very very hard) (Borg, 1982). Jump height (expressed in meters), jump height percentage (%) (expressed as the percentage of standing lower torso height), and stance time (expressed in seconds) were calculated by DARI's biomechanical analysis system during each performance analysis. Three repetitions were performed during each DARI analysis, and means were calculated for analysis. For our analysis, the presence of fatigue in participants was to be indicated by multiple of the following: Longer round times, cessation of exercise, progressive increase in stance time and decrease in jump height, significant increases in HR with decreased recovery in-between rounds, and significant increases in RPE.

Protocol

Interested individual were directed to complete a pre-participation screening survey (via Qualtrics™; Provo, UT), which collected information regarding health history, physical activity behaviors, diagnosed knee injuries/pathology, and current participation in HIFT programs. Participants who met the inclusion/exclusion criteria attended a familiarization session, which was used to obtain informed, written consent and image acknowledgement, assessments of required movement proficiencies, establish leg dominance, and to provide a full briefing on the testing protocol participants would undergo.

Table 1. Standardized Dynamic Warm-Up

Movement	*Repetitions/Duration*
Air assault bike	3 minutes
Forward overhead walking Lunge	5 meters
Backward overhead walking lunge	5 meters
Inch worms	5 meters
traveling adductor stretch	5 meters
Leg swings forward/backward	10 repetitions
Leg swings side/side	10 repetitions
Good mornings	15 repetitions
Air squats	15 repetitions

Participants were then scheduled for a testing session at the Orthopedic Sports Medicine Center (OSMC) in Manhattan, Kansas. Upon arrival, participants were fitted with a heart rate monitor and completed a standard dynamic warm-up (Table 1). This warm-up protocol was adapted from a standardized warm-up we have utilized previously (Cosgrove, Crawford, & Heinrich, 2019; Heinrich et al., 2015) and has demonstrated its effectiveness in preparing participants for exercise without inducing fatigue. Participants finished their warm-up performing each movement in the workout (i.e., dumbbell thruster, air squat, burpee box jump) three (3) times to demonstrate their readiness.

Next, participants entered the DARI system, where they performed each movement under DARI analysis a total of three (3) times in a randomized order. The drop vertical jump was performed by instructing participants, while standing on a wood box (24/20 inches for men and women, respectively), to step off the front as to drop down to ground and immediately transition into a maximum effort vertical jump. During the single-leg squat, participants were instructed to shift their weight completely to the test leg to a single-leg stance position. Participants then descended as low as possible, sending their foot posterior to the frontal plane, without touching the foot or knee to the ground. In order for the DARI analysis to be accurate, performance standards had to be met, meaning participants were asked redo the movement if it was executed incorrectly. For example, during the single-leg squat as the body descended if the free foot touched the ground at any point, the movement was repeated. When this occurred, the error was recorded, and participants repeated the movement until all repetitions were correctly performed. This served as the baseline biomechanical analysis data.

Participants then completed the first of three rounds of the HIFT workout (see Table 2). All participants completed the workout as prescribed, performing each repetition with no assistance or movement modifications. While each of the movements in the workout were chosen for their ability to induce fatigue in the lower extremity musculature, they were also chosen as movements that required knee joint stability and were regularly performed among HIFT participants. After completing the first round of the workout, participants completed a second DARI analysis, where they repeated the three movements in a randomized order, followed by a second round of the workout, third round of randomized order DARI analysis, third round of HIFT workout, and a fourth randomized order DARI analysis. In total, four rounds of DARI testing and three rounds of the workout were performed.

Table 2. Workout Protocol

3 rounds (each for time):
10 dumbbell (DB) thrusters
men: 25 lb. DB
women: 15 lb. DB
10 air-squats
10 burpee box jumps
men: 24 inch box
women: 20 inch box

Statistical Analysis

Statistical analyses were performed using the Statistical Package for the Social Sciences (SPSS) version 25.0 for Windows (IBM, Armonk, NY). Prior to inferential testing, all data were assessed for normality using both statistical (i.e., Shapiro-Wilk) and visual methods (i.e., Q-Q plots). Descriptive statistics were generated for all anthropometric data between sexes. Correlations were run for all anthropometric data in relation to primary study outcomes (e.g., knee valgus) to identify potential covariates. For specific aim 1, two-factor, repeated-measures [Sex (2) x Time (4)] MANOVAs were run to investigate potential main effects and interactions between men and women in fatigue parameters before, between, and after the workout rounds. For specific aim 2, one-way MANCOVAs (controlling for body mass and squat depth) were run to investigate baseline differences in kinematic and kinetic variables between men and women for each functional movement task (i.e., left and right single-leg squat; drop jump). For specific aim 3, two-factor repeated-measures [Sex (2) x Time (4)] MANCOVAs (controlling for body mass and loading depth) were run to investigate potential main effects and interactions in kinematic and kinetic variables between men and women for each functional movement task over time with men as the reference group. For all analyses, Bonferroni post hoc adjustments were used for pairwise comparisons and the alpha-level denoting statistical significance was set at $p \leq 0.05$.

Chapter 3 - Results

Participant Characteristics

Participants were healthy adults (10 men and 10 women) between the ages of 18-35 years (24.9 ± 4.7 years). Men were 179.8 ± 5.7 cm tall, weighed 84.2 ± 10.0 kg, and had a body fat percentage of 18.1 ± 5.2%. Women were 166.3 ± 5.3 cm tall, weighed 62.5 ± 8.5 kg, and had a body fat percentage of 25.8 ± 7%. Age did not differ significantly between men and women (24.4 ± 4.3 vs. 25.4 ± 5.4 years, $t = .50$, $p = .63$), although men were taller ($t = -5.50$, $p < .001$) and heavier ($t = -4.80$, $p < .001$). Conversely, women had a greater body fat percentage ($t = 2.80$, $p = .01$) than men. Nine of 10 of men and 9 of 10 women were right leg dominant.

Fatigue Parameters

Performance Time

On average, men (315.5 ± 95.8 s) completed the workout faster than women (342.2 ± 65.2 s), although no significant differences existed in mean round time between sexes (105.7 ± 16.8 s vs. 114 ± 16.8 s for men and women, respectively; mean diff. = 8.90 ± 23.90 s; 95% CI = -16.70 to 34.50 s; $F = .53$, $p = .48$, $\eta^2 = .03$). There was no significant main effect for time ($F = 3.60$, $p = .07$, $\eta^2 = .17$) or interaction for sex*time ($F = .01$, $p = .95$, $\eta^2 = 0$).

Heart Rate

As shown in Figure 1, there was a significant main effect of time on HR ($F = 213.31$, $p < .001$, $\eta^2 = .92$), exemplified by whole group changes in HR at each time point throughout the fatigue protocol. Participant HR increased significantly from start to finish of each round and decreased significantly between the end of each round and the start of the next. There were, however, no significant differences in HR between men and women at baseline or over time.

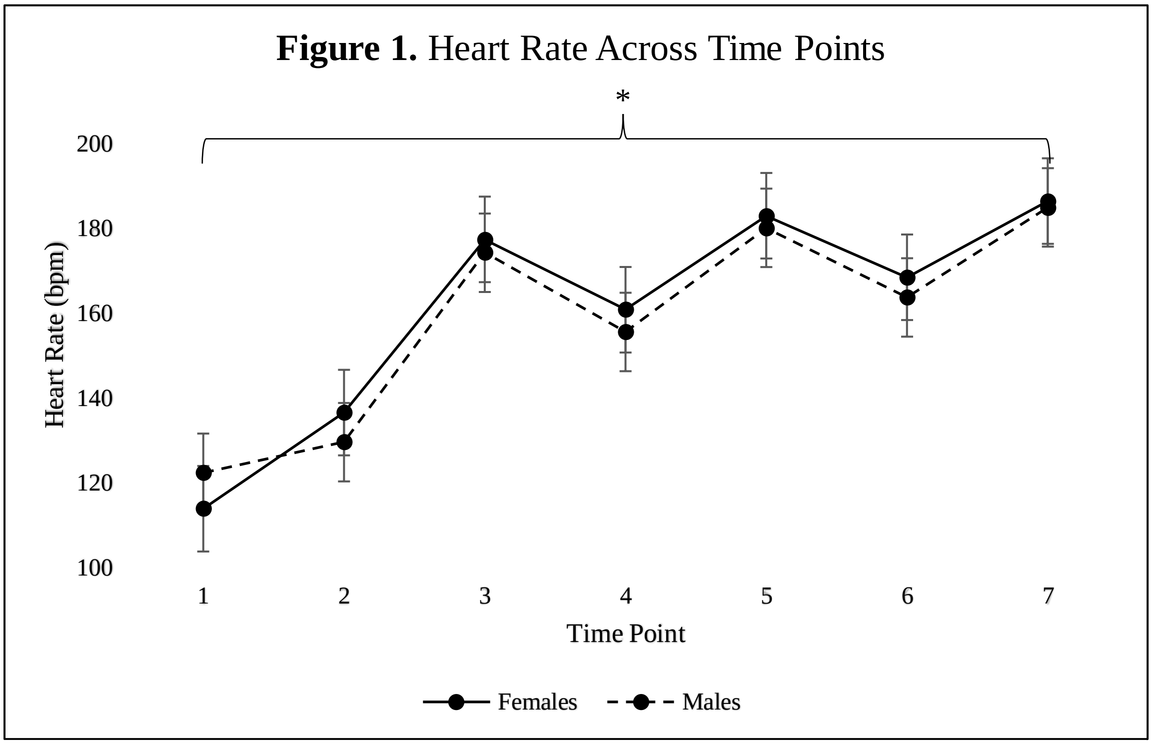


Figure 1. Participant HR over time during the workout. Time point 1 denotes the start of the protocol in the DARI system, and each subsequent time point denotes the start and finish of each round, respectively. * denotes significant whole group changes in HR.

Rate of Perceived Exertion

There was no significant sex*time interaction (F = .76, p = .55, η^2 = .04) for RPE during the workout. However, there was a significant univariate effect for time on whole group RPE (F = 22.75, p < .001, η^2 = .56). Specifically, whole group RPE increased each round of the workout.

Jump Height, Jump Height %, and Stance Time

While no sex, time, or sex*time-related differences occurred in stance time, there were significant univariate effects for sex on jump height in m (mean difference = -.13 ± .08; 95 CI = -.21 to -.04; F = 9.28, p < .01, η^2 = .34) and jump height % (mean difference = -.10 ± .08%; 95%

CI = -.19 to .01; $F = 5.53$, $p < .05$, $\eta^2 = .24$) at all time points. Additionally, there were no significant interaction between sex*time for stance time, jump height, or jump height %.

Table 3. Comparisons in Fatigue Parameters Between Men and Women

Fatigue2 Parameters	Men (n = 10)	Women (n = 10)	Main Effect (Time)	Main Effect (Sex)	Sex x Time Interaction
Total Time (sec)			.07	.48	.95
Round 1	97.6 ± 17.9	106.7 ± 11.7			
Round 2	103.8 ± 31.3	113.3 ± 22.3			
Round 3	114.1 ± 51.8	122.2 ± 36.0			
HR (bpm)			< .001*	.74	.06
Start of Protocol	122.4 ± 27.1	114 ± 21.1			
Start Round 1	129.7 ± 17.1	136.6 ± 17.5			
End Round 1	174.3 ± 13.7	177.5 ± 11.0			
Start Round 2	155.7 ± 21.3	160.9 ± 14.8			
End Round 2	180.2 ± 12.8	183.1 ± 10.6			
Start Round 3	163.8 ± 18.3	168.6 ± 14.0			
End Round 3	185.0 ± 11.7	185 ± 11.7			
RPE (6-20)			<.001*	.17	.26
Round 1	13.9 ± 2.1	15.5 ± 1.8			
Round 2	15.8 ± 1.7	17.0 ± 2.1			
Round 3	17.5 ± 2.0	17.7 ± 1.6			
Jump Height (m)			.08	< .01#	.92
Time Point 1	.49 ± .1	.37 ± .1			
Time Point 2	.49 ± .1	.36 ± .1			
Time Point 3	.48 ± .1	.35 ± .1			
Time Point 4	47 ± .1	.34 ± .1			
Jump Height %			.08	.03#	.88
Time Point 1	50.6 ± 11.9	41.3 ± 8.6			
Time Point 2	49.9 ± 10.1	39.8 ± 6.8			
Time Point 3	49.5 ± 10.7	39.0 ± 8.7			
Time Point 4	48.4 ± 12.5	38.3 ± 8.6			
Stance Time (sec)			.39	.57	.44
Time Point 1	.37 ± 3	.39 ± .1			
Time Point 2	.41 ± .2	.37 ± .1			
Time Point 3	.38 ± .1	.35 ± .1			
Time Point 4	.38 ± .2	.3 ± .2			

Raw mean scores and supporting statistical data for all fatigue-related main effects and interactions.
*all time points significantly different from one another, # significant difference between men and women

Ground Reaction Forces (GRFs)

There was a significant univariate main effect of sex for right leg GRFs in Newtons (mean difference = -140.6 ± 115.6; 95% CI = -256.9 to -24.3; F = 5.68, p = .02, η^2 = .02) at peak loading, but not the left (F = 1.23. p = .27, η^2 = .01). However, no significant interactions between sex*time for GRFs were detected at peak loading or takeoff. Raw mean scores and supporting statistical data for main effects and interactions are found in Table 4**.**

Table 4. Sex Comparisons in Ground Reaction Forces (N) at Peak Loading and Takeoff

Time Points (Newtons)	Men (n = 10)	Women (n = 10)	Main Effect (Time)	Main Effect (Sex)	Sex x Time Interaction
Right Leg Peak Loading			.36	.02#	.85
Time Point 1	1551.1 ± 424.7	1113.6 ± 360.7			
Time Point 2	1604.4 ± 471.7	1221.2 ± 393.2			
Time Point 3	1655.1 ± 405.0	1258.0 ± 405.7			
Time Point 4	1593.8 ± 365.4	1180.0 ± 435.3			
Right Leg Take-off			.25	.24	.85
Time Point 1	1469.6 ± 437.7	1089.0 ± 429.9			
Time Point 2	1561.1 ± 525.3	1236.4 ± 443.8			
Time Point 3	1616.6 ± 525.3	1253.4 ± 440.0			
Time Point 4	1534.2 ± 471.0	1175.2 ± 468.8			
Left Leg Peak Loading			.06	.27	.34
Time Point 1	1356.1 ± 371.4	1027.2 ± 268.6			
Time Point 2	1486.3 ± 384.7	1162.1 ± 240.2			
Time Point 3	1578.9 ± 460.2	1150.5 ± 324.3			
Time Point 4	1591.2 ± 473.8	1110.6 ± 332.6			
Left Leg Take-off			.13	.36	.34
Time Point 1	1294.4 ± 394.3	972.1 ± 341.5			
Time Point 2	1427.9 ± 419.0	1156.5 ± 309.3			
Time Point 3	1517.9 ± 534.4	1078.1 ± 365.1			
Time Point 4	1512.7 ± 517.8	1030.2 ± 386.9			

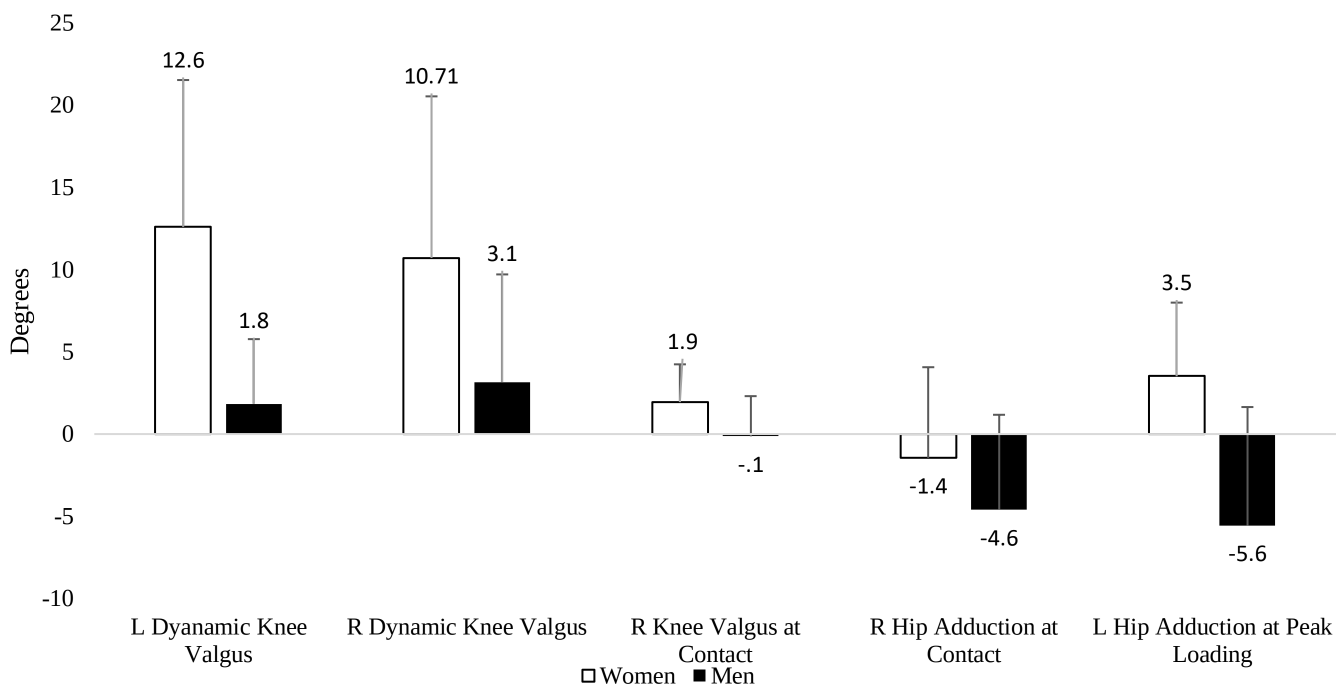
significant difference between men and women

Joint Kinematics and Kinetics

Drop Vertical Jump

There were significant univariate main effects for sex on baseline drop vertical jump kinematics and kinetic variables. For the left leg: dynamic knee valgus (mean difference = 12.2 ± 5.3 degrees; 95% CI = 6.8 to 17.6; F = 20.35, p < .001, η^2 = .27),), and hip adduction at peak loading (mean difference = 7.2 ± 4.7; 95% CI = 2.4 to 12.0, F = 9.09, p > .01, η^2 = .14). For the right leg: dynamic knee valgus (mean difference = 12.7 ± 6.5; 95% CI = 6.0 to 19.3; F = 14.59, p < .001, η^2 = .21), hip adduction at contact (mean difference = 5.8 ± 4.3; 95% CI = 1.4 to 10.1; F = 7.07, p = .01, η^2 = .11), knee valgus at contact (mean difference = 2.4 ± 1.9; 95% CI = .5 to 4.4; F = 6.13, p = .016, η^2 = .10), and knee flexion at peak loading (mean difference = 3.9 ± 3.6; 95% CI = .3 to 7.6; p = .04, η^2 = .08). Figure 2 displays these significant differences between men and women during the drop vertical jump at baseline.

Figure 2. Sex Differences in Drop Vertical Jump Kinematics at Baseline



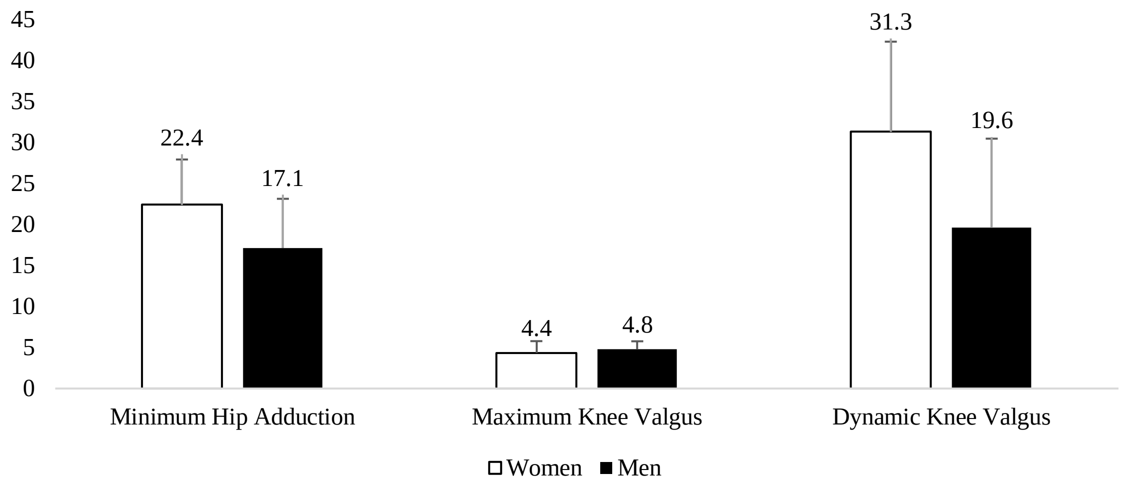
Left Single-Leg Squat

No multivariate main effect for sex was present at baseline for the left single-leg squat variables of interest (F = 1.29, p = .26, η^2 = .26).

Right Single-Leg Squat

There was a significant univariate effect for sex at baseline for the following variables: minimum hip adduction (mean difference = 8.5± 4.5; 95% CI = 4.0 to 13.0; F = 13.95, p < .001, η^2 = .20), maximum knee valgus (mean difference -1.4 ± .9; 95% CI = -2.3 to -.5; F = 8.86, p < .01, η^2 = .14), and dynamic knee valgus (mean difference = 20.1 ± 8.3; 95% CI = 11.6 to 28.6; F = 22.66, p < .001, η^2 = .29). Sex differences in knee mechanics during the right single-leg squat are outlined in Figure 3.

Figure 3. Sex Differences in Right Single-Leg Squat at Baseline



Kinematics and Kinetics over Time

Drop Vertical Jump

There was a significant univariate effect for sex across all time points for kinematic and kinetic variables in both legs. In the right leg, the following variables were significantly different across all time-points between men and women: hip adduction at contact (mean difference = 3.4± 1.7; 95% CI = 1.8 to 5.1; F = 16.83, p < .001, η^2 = .07), dynamic knee valgus (mean difference = 6.7 ± 2.7; 95% CI = 4.0 to 9.4; F = 23.38, p < .001, η^2 = .09), hip flexion at peak loading (mean difference = 13.4 ± 5.5; 95% CI = 7.7 to 19.1; F = 21.50, p < .001, η^2 = .09), knee torque at peak loading (mean difference = -66.6 ± 27.8; 95% CI = -94.6 to -38.6; F = 22.02, p < .001, η^2 = .09), and knee torque at takeoff (mean difference = -56.5 ± 29.4; 95% CI = -86.0 to -26.9; F = 14.17, p < .001, η^2 = .06). In the left leg, the following variables were significantly different across all time-points between men and women: hip adduction at contact (mean difference = 2.9 ± 1.7; 95% CI = 1.1 to 4.6; F = 10.78; p < .001, η^2 = .05), maximum knee valgus (mean difference = -1.1 ± 1.1; 95% CI = -2.3 to -.1; F = 4.82, p = .03, η^2 = .02), dynamic knee valgus (mean difference = 8.5 ± 3.2; 95% CI = 5.3 to 11.7; F = 27.14, p < .001, η^2 = .12), hip flexion at peak loading (mean difference = 12.4 ± 5.6; 95% CI = 6.8 to 18.0; F = 19.05, p < .001 η^2 = .08), hip adduction at peak loading (mean difference = 4.6 ± 2.7; 95% CI = 1.9 to 7.2; F = 11.49, p = .001, η^2 = .05). There was no interaction between sex*time (right leg: F = .76, p = .85, η^2 = .04; left leg: F = 1.08, p = .34, η^2 = .06). Table 6 compares raw mean scores during the drop vertical jump for men and women by time and sex across all time points.

Table 5. Sex Comparisons in Drop Vertical Jump Kinematics and Kinetics by Time and Sex Across all Time Points

Kinematic Parameters (degree)	Men (n = 10)	Women (n = 10)	Main Effect (Time)	Main Effect (Sex)	Men (n = 10)	Women (n = 10)	Main Effect (Time)	Main Effect (Sex)
	Right Leg				*Left Leg*			
Hip Flexion at Contact			.58	.15			.43	.58
Time Point 1	3.1± 8.1	15.7 ± 14.7			4.6 ± 9.2	13.6 ± 8.6		
Time Point 2	5.1 ± 9.3	11.7 ± 7.9			7.3 ± 8.9	12.5 ± 7.8		
Time Point 3	7.6 ± 9.3	13.2 ± 11.4			9.3 ± 7.8	13.9 ± 10.9		
Time Point 4	6.49 ± 7.5	10.7 ± 11.9			7.8 ± 7.3	11.8 ± 11.3		
Hip Adduction at			.68	< .001#			.76	< .01#
Time Point 1	-4.6 ± 5.8	-1.5 ± 5.5			-3.2 ± 4.3	-.8 ± 7.7		
Time Point 2	-3.3 ± 4.9	-.8 ± 3.6			-3.2 ± 4.1	-.0 ± 3.5		
Time Point 3	-3.3 ± 4.1	-.0 ± 3.3			-3.0 ± 4.1	.6 ± 3.4		
Time Point 4	-4.7 ± 5.1	.5 ± 3.7			-2.7 ± 3.6	.6 ± 2.9		
Knee Flexion at			.31	.73			.05	.10
Time Point 1	25.5 ± 15.6	35.5 ± 23.0			30.6 ± 19.2	32.6 ± 21.1		
Time Point 2	20.9 ± 12.5	20.4 ± 15.8			24.8 ± 14.0	20.7 ± 16.8		
Time Point 3	21.1 ± 11.0	29.2 ± 22.6			24.6 ± 12.9	28.0 ± 23.3		
Time Point 4	24.5 ± 12.3	29.6 ± 20.3			26.1 ± 13.1	30.7 ± 19.8		
Knee Valgus at Contact			.23	.07			.59	.34
Time Point 1	-.1 ± 2.4	1.9 ± 2.3			-.6 ± 3.1	-.8 ± 2.5		
Time Point 2	-.3 ± 2.7	.9 ± 2.7			-.7 ± 3.3	-1.3 ± 2.3		
Time Point 3	-.8 ± 2.5	1.4 ± 3.2			-1.0 ± 2.5	-1.0 ± 1.9		
Time Point 4	-.4 ± 2.5	1.0 ± 2.6			.1 ± 2.9	-1.0 ± 2.5		
Maximum Knee Valgus			.40	.27			.55	.03#
Time Point 1	1.2 ± 2.8	3.7 ± 1.9			.2 ± 3.0	.3 ± 2.7		
Time Point 2	1.0 ± 2.8	2.9 ± 2.8			.7 ± 3.0	.2 ± 2.6		
Time Point 3	.7 ± 2.4	3.5 ± 3.0			.0 ± 2.4	.2 ± 2.8		
Time Point 4	1.5 ± 3.1	2.6 ± 2.6			1.4 ± 2.8	.1 ± 2.9		
Dynamic Knee Valgus			.85	<.001#			.46	< .001
Time Point 1	3.2 ± 6.6	10.7 ± 9.8			1.8 ± 3.9	12.6 ± 8.9		
Time Point 2	4.5 ± 6.0	11.1 ± 6.0			3.8 ± 6.3	15.1 ± 12.2		
Time Point 3	4.4 ± 6.5	10.2 ± 5.7			4.0 ± 7.4	12.8 ± 7.5		
Time Point 4	4.4 ± 8.8	9.2 ± 5.4			3.7 ± 6.3	12.6 ± 11.2		
Knee Torque at			.45	< .001#			.39	< .01#
Time Point 1	254.6 ± 95.0	167.4 ± 68.9			203.2 ± 68.4	142.2 ± 56.2		
Time Point 2	348.6 ± 80.0	165.7 ± 56.6			221.9 ± 72.8	157.7 ± 50.7		
Time Point 3	257.0 ± 96.0	172.1 ± 71.2			243.5 ± 85.6	151.5 ± 65.4		
Time Point 4	242.6 ± 77.8	157.6 ± 68.3			240.5 ± 88.7	146.9 ± 55.8		
Knee Torque at Takeoff			.31	< .001#			.63	< .01#
Time Point 1	242.7 ± 96.9	164.8 ± 76.1			193.2 ± 77.1	137.6 ± 62.4		
Time Point 2	248.8 ± 94.3	177.5 ± 75.9			213.3 ± 88.2	159.2 ± 64.6		
Time Point 3	264.3 ±	178.0 ± 85.5			241.4 ±	137.7 ± 74.7		
Time Point 4	249.4 ± 85.6	158.9 ± 76.5			232.8 ±	136.6 ± 54.7		

significant difference between men and women

Right Single-Leg Squat

There was a significant univariate effect for sex across all time points for the following variables: minimum hip adduction (mean difference = 5.6± 2.2; 95% CI = 3.4 to 7.8; F = 24.52, p < .001, η^2 = .10), maximum knee valgus (mean difference = -1.3 ± .5; 95% CI = -1.8 to -.8; F = 29.35, p < .001, η^2 = .11), and dynamic knee valgus (mean difference = 11.9 ± 4.0; 95% CI = 7.8 to 15.9; F = 33.21, p < .001, η^2 = .13). No significant interaction between sex*time (F = 1.08, p = .35, η^2 = .04) was detected for any right single-leg squat variables.

Left Single-Leg Squat

There was a significant univariate effect for sex across all time points for the following variables: minimum hip adduction (mean difference = -2.2± 2.1; 95% CI = -4.4 to -.1; F = 4.21, p = .04, η^2 = .02), maximum hip adduction (mean difference = -1.2 ± 1.2; 95% CI = -2.4 to -.03; F = 4.07, p = .05, η^2 = .02), maximum hip flexion (mean difference = 10.2 ± 5.1; 95% CI = 5.1 to 15.2; F = 15.45, p < .001, η^2 = .06), hip flexion at loading (mean difference = 10.5 ± 5.2; 95% CI = 5.3 to 15.7; F = 15.86, p < .001, η^2 = .02), maximum knee valgus (mean difference = -1.7 ± .7; 95% CI = -2.4 to -1.0; F = 22.07, p < .001, η^2 = .09), and knee valgus at loading (mean difference = -2.5 ± 1.1; 95% CI = -3.6 to -1.4; F = 1.63, p < .001, η^2 = .01). No interaction between sex*time (F = 0.60, p = .97, η^2 = .03) was detected either.

Table 6. Sex Comparisons in Single-Leg Squat Kinematics and Kinetics by Time and Sex Across all Time Points

Kinematic Parameters	Men (n = 10)	Women (n = 10)	Main Effect (Sex)	Men (n = 10)	Women (n = 10)	Main Effect (Sex)
	Right Leg			*Left Leg*		
Minimum Hip Adduction			< .001#			.04#
Time Point 1 (deg)	17.1 ± 6.0	22.4 ± 5.5		14.4 ± 5.1	17.7 ± 5.0	
Time Point 2 (deg)	18.3 ± 4.9	22.4 ± 4.8		15.4 ± 5.1	17.2 ± 7.6	
Time Point 3 (deg)	18.0 ± 4.7	21.2 ± 5.3		15.6 ± 4.0	17.5 ± 6.2	
Time Point 4 (deg)	18.1 ± 4.8	22.1 ± 7.4		14.8 ± 5.9	17.5 ± 6.3	
Maximum Hip Adduction			.18			.05#
Time Point 1 (deg)	-2.3 ± 3.3	-.7 ± 2.0		-2.4 ± 3.5	-1.4 ± 3.5	
Time Point 2 (deg)	-1.7 ± 2.5	-1.5 ± 2.6		-2.7 ± 3.8	-2.0 ± 4.0	
Time Point 3 (deg)	-2.5 ± 4.8	-.5 ± 1.3		-1.9 ± 2.4	-.7 ± 2.1	
Time Point 4 (deg)	-2.1 ± 3.2	-1.0 ± 2.2		-2.3 ± 3.3	-.7 ± 1.7	
Maximum Knee Valgus			<.001#			<.001#
Time Point 1 (deg)	4.8 ± 1.0	4.4 ± 1.4		3.8 ± 1.8	3.3 ± 2.0	
Time Point 2 (deg)	4.5 ± 1.1	4.8 ± 1.2		3.8 ± 1.6	3.4 ± 2.1	
Time Point 3 (deg)	4.6 ± 1.5	4.7 ± 1.4		3.5 ± 1.9	3.7 ± 2.1	
Time Point 4 (deg)	4.4 ± 1.6	4.6 ± 1.4		3.3 ± 2.0	3.2 ± 2.3	
Knee Valgus at Loading			.30			<.001#
Time Point 1 (deg)	2.2 ± 3.0	1.0 ± 2.8		.5 ± 3.0	-.6 ± 2.6	
Time Point 2 (deg)	1.1 ± 3.4	2.5 ± 2.8		.9 ± 3.4	-.9 ± 2.3	
Time Point 3 (deg)	1.4 ± 3.1	2.5 ± 3.4		.8 ± 3.1	-.3 ± 2.6	
Time Point 4 (deg)	1.4 ± 3.3	2.1 ± 3.3		. ± 3.2	-1.2 ± 2.4	
Dynamic Knee Valgus			<.001#			.203
Time Point 1 (deg)	19.6 ± 10.9	31.3 ± 17.0		15.4 ± 8.2	20.5 ± 9.0	
Time Point 2 (deg)	22.3 ± 10.1	28.7 ± 8.3		17.2 ± 9.5	19.3 ± 10.6	
Time Point 3 (deg)	21.5 ± 10.1	27.3 ± 10.2		17.3 ± 6.7	19.6 ± 11.4	
Time Point 4 (deg)	21.6 ± 10.2	27.3 ± 10.5		15.0 ± 9.8	19.0 ± 10.7	
Maximum Hip Flexion			.03#			<.001#
Time Point 1 (deg)	95.2 ± 17.5	95.4 ± 15.4		96.3 ± 13.6	97.6 ± 15.9	
Time Point 2 (deg)	96.2 ± 17.6	92.4 ± 12.9		96.6 ± 14.7	95.0 ± 16.9	
Time Point 3 (deg)	95.5 ± 16.1	93.3 ± 16.7		95.7 ± 14.8	94.6 ± 16.1	
Time Point 4 (deg)	93.7 ± 17.8	88.4 ± 18.2		93.6 ± 15.8	92.4 ± 17.6	
Hip Flexion at Loading			.04#			<.001#
Time Point 1 (deg)	94.2 ± 17.9	94.1 ± 15.8		95.0 ± 14.4	96.6 ± 15.9	
Time Point 2 (deg)	95.6 ± 18.1	91.2 ± 12.9		95.7 ± 15.1	94.1 ± 16.8	
Time Point 3 (deg)	94.9 ± 16.1	91.7 ± 17.2		95.0 ± 15.3	93.7 ± 16.7	
Time Point 4 (deg)	93.2 ± 17.9	86.7 ± 18.3		92.9 ± 16.2	91.4 ± 18.7	

significant difference between men and women

Chapter 4 - Discussion

Aim 1.

The first aim of this study was to determine whether a workout incorporating traditional HIFT movement modalities and appropriate time standards caused significant fatigue in experienced HIFT men and women. Participant fatigue was evaluated by measuring performance, physiological, and perceptual markers throughout the workout. While each of these variables played a role in determining the level to which an individual was experiencing fatigue, a wholistic approach was necessary for interpreting whether fatigue actually occurred.

Our hypotheses were two-fold. First, we predicted the workout would cause significant fatigue in both men and women. Second, we predicted that men and women would both display similar physiological (i.e., heart rate) and perceptual (i.e., rating of perceived exertion) responses to exercise. Our results did not indicate that participants experienced significant fatigue, but men and women did display similar physiological and perceptual responses to the workout.

Fatigue Parameters.

Participants' abilities to maintain jump height, jump height %, and stance time across all time-points suggested no deterioration in performance, even though they perceived themselves to be working harder as the workout went on. These data are important to consider in the context of our findings as they do not clearly show fatigue as a function of the inability to maintain work (Green, 2010).

We utilized participant stance time data to objectively determine whether participants were able to maintain power output throughout the workout. Since participants maintained vertical jump height, they also maintained RFP and power output. These data show no indication of fatigue being present during the workout.

Physical and Perceptual Reponses.

Men averaged faster completion times for each round of the workout as well as the entire workout than women, although these differences were not statistically significant. Both sexes displayed expected HR responses to the workout, with significant linear increases in HR as a result of increased work rate during each subsequent round, and linear decreases between rounds when the stimulus was lower. HIFT workouts are designed to constantly vary in time and modes used, but because they are performed at relative high intensity, there is generally little time for recovery between movements (Leyland, 2007). For the purposes of our study we had to design a workout that had built in intervals between each round when performance testing could take place. Because participants displayed increasing start and finish HRs progressively throughout the workout, we determined that less recovery was taking place after each round and thus the workout successfully kept participant HR elevated, even with short intervals in-between each round.

Whole group RPE increased significantly across all three time points, and did not differ by sex. Thus, women perceived the intensity of the workout the same as men. RPE is an important component of determining fatigue, as it quantifies the sensation of fatigue as a conscious perception of increasing effort needed to sustain submaximal exercise (Gibson et al., 2003). Whole-body RPE integrates awareness of sensations arising from the muscle, joints, chest, skin, circulating factors, and inputs from higher brain centers (Knicker et al., 2011), making it perhaps the best measure attainable in determining perceived exertion occurring during a workout. However, while RPE is a strong predictor of HR (Dunbar et al., 1992), because it is a measure of perceived exertion, it is not synonymous with fatigue. The RPE trends of these data are what we expected to see, and further supported the notion that perceived work rate increased significantly throughout the workout. However, because these performance data provided no indication that work could not be maintained, we have no definitive indication that RPE alone confirmed that fatigue occurred.

Further supporting this finding were the performance measures evaluated during the workout. First, no significant differences in round time occurred between sexes at any time point or over time. This meant that not only did both groups perform similarly during the workout, but also that they did not lose the ability to complete the prescribed work as quickly as possible. Reasons for this may have included the used of relatively simple movements, lighter weight dumbbells, and round times and repetition schemes that were short enough in duration that participants may not have had sufficient time to experience the feeling of fatigue during the workout. Longer rounds or more repetitions may have caused participants to slow their work rate over the course of several minutes.

Aim 2.

The second aim of this study was to determine whether sex differences in movement mechanics existed at baseline. We hypothesized that women would exhibit movement patterns more consistent with known predictors for non-contact ACL injury than men at baseline during both the drop vertical jump and the single-leg squat. Consistent with this hypothesis, women's knee and hip mechanics were significantly worse during the drop vertical jump and right single-leg squat than men. Women did perform as well as men during the left single-leg squat, but this was somewhat expected as the left leg was the stabilizing leg for 90% of both men and women.

Drop Vertical Jump.

There was a significant main effect for sex during baseline drop vertical jump performance between men and women. Specifically, women exhibited significantly greater dynamic knee valgus in both knees than men. This finding was consistent with the current literature and indicates that women exhibited knee mechanics consistent with increased risk for acute non-contact knee injury (F. R. Ford et al., 2003; Kernozek et al., 2005, 2008). Dynamic knee valgus is a strong indicator of dynamic knee stability during the drop vertical jump as it considers not only peak valgus moments at the knee, but change from smallest to largest as well. Generally speaking, this is where we see a demonstrated failure to stabilize the knee during dynamic movement such as a drop landing.

Women also landed with greater right hip adduction and knee valgus at contact, as well as greater left hip adduction and right knee flexion at peak loading than men. Greater knee flexion at peak loading generally occurs in concurrence with more erect torso posture. This condition results in translation of force to the anterior knee, placing the ACL under greater stress (Dingenen et al., 2015). High knee flexion angles along with significant valgus moments on the knee place the joint in a much more vulnerable state (Baldon et al., 2009).

Left Single-Leg Squat.

Interestingly, no differences were detected at baseline for the left single-leg squat. Previous research has demonstrated that, compared to the dominant limb, the non-dominant limb has a more effective protective mechanism because of its ability to effectively restrain excess joint motion (Niu et al., 2011). Nine of 10 of men and 9 of 10 women were right leg dominant. As the non-dominant leg is the stabilizer for high impact movements such as kicking a ball, this would predict left leg stabilization to be highly proficient, despite it being the non-dominant limb for 90% of participants.

Right Single-Leg Squat.

Baseline data suggested that women had less dynamic control of their dominant limb than men. To this point, McCurdy & Langford (2005) compared dominant to non-dominant limb strength in healthy young men and women using a modified unilateral squat protocol that allowed for stabilization of the non-test leg to remove balance as a limiting factor. By accounting solely for unilateral strength, they found no side-to-side strength differences in either sex, which lends further support that limb strength plays a key role in dynamic stabilization of the knee joint during single-leg squats.

Aim 3.

The third aim of this study was to determine whether sex was associated with changes in dynamic knee stability across the exercise protocol. Although we expected women to exhibit less knee stability than men at baseline, we did not expect their knee stability to deteriorate significantly more during the workout protocol. Our data supported this hypothesis in that no significant interactions between sex*time were detected for any biomechanical measures.

Drop Vertical Jump.

There were significant differences in knee joint kinetics and kinematics between men and women sex during the drop jump across all time points. Women demonstrated less knee stability during the drop vertical jump, but their movement did not deteriorate over time. Specifically, women landed with significantly greater hip adduction at contact than men. Additionally, women consistently had significantly greater right and left knee dynamic valgus and maximum left knee valgus during the drop vertical jump.

Further, at peak loading, women achieved greater hip flexion angles, but fell into significant greater left hip adduction. While high levels of hip adduction are not ideal during peak loading, increased hip flexion at peak loading is indicative of safer and more efficient movement (Dingenen et al., 2015). Proper loading of the posterior chain keeps the body's center of mass in line with the frontal plane, which minimizes stress placed on the knee during flexion.

Left Single-Leg Squat

There were, again, significant differences in knee joint mechanics between men and women, although in this instance, women actually performed better. Women exceeded expectations during the left single-leg squat, contrary to what is accepted in the literature about movement in women. These results contradict the notion that women have been found to naturally perform the single-leg squat with greater medial deviation at the knee and internal rotation of the femur caused by joint actions at the hip (Kianifar et al., 2017).

In this case, women exhibited greater maximum hip flexion as well as greater hip flexion at max loading. This is indicative of strong loading mechanics and is likely directly related to their superior knee mechanics across all time-points for the left single-leg squat compared to men. Previous research has indicated that moderate forward trunk leans of approximately 40 degrees may lower non-contact ACL strains and increase activation of the hamstrings while decreasing activation of the quadriceps muscles, preventing anterior tibial translation (Kianifar et al., 2017). Men exhibited greater minimum and maximum hip adduction, maximum knee valgus, and greater knee valgus at low point than women.

Right Single-Leg Squat.

Right single-leg squat performance reflected that of baseline testing in both sexes. Women had significantly greater minimum hip adduction, maximum knee valgus, and dynamic knee valgus than men. These results suggest some interesting possibilities as to the mechanisms that may account for men having significantly better knee mechanics on their dominant limb than women. First, because limb-to-limb strength is relatively symmetrical in healthy men and women

(McCurdy & Langford, 2005), muscular strength likely plays an stabilizing role in the dominant leg. However, neuromuscular control may be a more important indicator of dynamic control of the dominant leg, as it relates to the ability to resist valgus knee moments and lateral trunk displacement (Hall et al., 2015).

Some aspects of our findings are consistent with those of Weeks et al. (2015), who also studied sex differences in single-leg squat performance before and after a lunge fatigue protocol in healthy men and women. Weeks found no interactions between time and sex on knee mechanics during the single-leg squat. Further, their protocol design ensured objective fatigue could be measured in all participants in that it required cessation of exercise or the inability to jump 80% of their max vertical jump. Thus, while we were unable link participant fatigue with changes in movement patterns, Weeks et al. (2015) reported the same findings with objective fatigue present in their participants.

Study Considerations

We found that while women generally exhibited knee mechanics associated with increased risk for acute non-contact knee injury, they were able to maintain movement fidelity throughout a HIFT workout. Moreover, while both men and women demonstrated high levels of movement proficiency, women actually exhibited more efficient mechanics during the left single-leg squat throughout the workout than men did. These findings are significant in that they suggest that HIFT participants may exhibit safer and more advantageous movement patterns than participants of similar non-contact sports. Criticisms have been made that due to high movement volumes and workout intensities, deterioration of movement mechanics likely lead to injuries among HIFT participants (Drum et al., 2016; Greely, 2014; Powers et al., 2014). Conversely, we found that while participants did perceive elevated levels of exertion, knee mechanics and overall performance did not deteriorate throughout the workout. This has clear injury preventation implications as these HIFT participants' movement mechanics did not significantly deteriorate over time.

There were several limitations to this study that warrant acknowledgement. Investigating whether sex differences existed in knee mechanics as a result of progressive fatigue during exercise (Aim 3) was contingent on participants being fatigued. Yet, because fatigue was not determined to have occurred (Aim 1), we were unable to extrapolate our findings directly to the effects of fatigue on lower extremity mechanics. A similar study (Weeks et al., 2015) used a protocol designed to quantify fatigue as either cessation of exercise (failure to continue) or the inability to maintain 80% of a previously established vertical jump max. This design held a specific definition for fatigue and ensured that their results were in context of fatigue being present. Further, while participants in this study were exposed to a workout which incorporated traditional HIFT methodology, the nature of constant variation in exercise duration and modalities meant that this workout was not all-encompassing in its representation of HIFT workouts. Workouts using greater loads, increased time-under-tension, more repetitions, or longer durations could all play roles in inducing fatigue. Finally, participants in this study were not only healthy young adults, but HIFT-trained with experience and instruction in correct movement, especially at high intensities. Untrained individuals, as well as those who do not regularly participate in high intensity exercise may have responded differently to this type of stimulus.

Chapter 5 - Conclusion

In this study of changes in movement mechanics during a HIFT workout, participants perceived high levels of exertion, but did not experience clearly defined fatigue during the workout. Women exhibited movement kinetics and kinematics more closely tied to increased risk of non-contact ACL injury during the drop vertical jump at baseline than men. These trends were similar during the single leg squat in the dominant leg, but not in the non-dominant leg, as no sex-differences were detected at baseline. On the contrary, women exhibited superior knee mechanics in the non-dominant leg throughout the workout, which to the best of our knowledge, differs from the existing literature. Finally, we found no significant decline or interaction between sex and changes in knee biomechanics throughout the workout. However, since significant fatigue was not substantiated in response to the workout, we cannot definitively state whether fatigue might alter knee biomechanics during a HIFT workout. Although women did display some mechanics associated with increased risk for non-contact ACL injury, those mechanics did not deteriorate during the workout. A myriad of reasons could explain why HIFT-trained women demonstrated high levels of movement fidelity, but it was likely a combination of constant coaching participants receive during workouts, proper on-ramp programs with appropriate exposure and introduction to greater levels of intensity over time.

Directions for Future Research

To better understand the fatiguing effects of HIFT workouts on the body, similar studies using objective measures of fatigue such as failing to maintain 80% of maximum jump height, or failure to maintain work rate should be conducted with HIFT populations. Fatigue protocols should use mainly closed kinetic chain movement that are repeatable and may be performed until participants reach exhaustion or fail to maintain adequate work rate such that fatigue may be measured objectively. If these findings can be replicated in experienced HIFT participants, longitudinal studies should be conducted on new HIFT participants to track progress over time to determine whether HIFT participation plays a direct role in improving movement patterns and lower limb stability.

Chapter 6: References

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